Effects of contact stress on patellarfemoral joint and quadriceps force in fixed and mobile-bearing medial unicompartmental knee arthroplasty

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Abstract

Background Unicompartmental knee arthroplasty (UKA) is an effective treatment for end-stage, symptomatic unicompartmental osteoarthritis (OA) of the knee joint. However, patellofemoral (PF) joint degeneration is a contraindication to medial UKA. Therefore, the objective of this study is to evaluate the biomechanical effect on the PF joint in medial UKA using fixed-bearing (FB) and mobile-bearing (MB) design prostheses.

Methods A three-dimensional finite-element model of a normal knee joint was developed using medical image data. We performed statistical analysis for each model. The differences in the contact stress on the PF joint and the quadriceps force between the FB and MB designs were evaluated under a deep-knee-bend condition.

Results At an early flexion angle, the results of the contact stress were showed that there was no significant difference between the FB and MB medial UKA models compared with the intact model. However, at a large flexion angle, we observed a significant increase in the contact stress of FB models. On the contrary, in the case of the MB models, there was no statistically significant increment compared to the intact model. Our results indicate that with medial UKA, the contact stress increased, and a greater quadriceps force was applied to the PF joint. However, there should be no difficulty in performing UKA on a PF joint with OA, unless there is anterior knee pain. This is because the increase in the contact stress is negligible.

Conclusions Our results showed that there was no significant difference in contact stress on the PF joint between medial UKA and intact knee joints. In particular, such a mechanism was easily found in mobile-bearing medial UKA. Therefore, this study biomechanically showed that degenerative changes in the PF joint should not be considered an absolute contraindication for treatment with medial UKA.

Background

Unicompartmental knee arthroplasty (UKA) is a surgical treatment alternative to total knee arthroplasty for isolated medial compartmental arthritis of a knee joint. The benefits of UKA include fewer complications, faster recovery, improved functional outcomes, and cost-effectiveness (1–4). Therefore, medial UKA has been increasingly used for the treatment of medial compartmental osteoarthritis (OA) over the past two decades (5). Historically, patellofemoral (PF) joint degeneration—and more specifically, advanced lateral PF joint facet degeneration—along with anterior knee pain, has been considered as an exclusion criterion for medial UKA (6, 7). However, it has recently been reported that PF joint degeneration does not influence clinical outcomes in UKA (8, 9). Additionally, there exists controversy regarding whether pre-existing PF joint degeneration is a contraindication for the performance of UKA. Thein et al. recently performed a study to determine the effect of medial fixed-bearing (FB) UKA on postoperative PF joint congruence and analyzed the effect of preoperative PF joint degeneration on the clinical outcome (10). No correlation was observed between the preoperative PF joint congruence or degeneration severity and the WOMAC scores at two-year follow-up (10). Preoperative PF joint congruence and degenerative changes do not affect UKA clinical outcomes (10). However, multiple studies using the Oxford knee system indicated that neither preoperative anterior knee pain nor moderate radiological PF osteoarthritic changes affected long-term clinical outcomes and survivorship for mobile-bearing (MB) UKA (9, 11, 12). One study suggested that MB UKA provides better restoration of normal knee kinematics, which theoretically translates to better patella tracking and long-term outcomes (13). Although several studies have revealed no significant difference in the clinical outcomes and complication rates between FB and MB UKA designs, the mode of failure often differs (14). Additionally, there has been a lack of research on the biomechanical effect on the PF joint with medial UKA. The biomechanical effect on the PF joint can be investigated via evaluation of the contact stress and quadriceps force after medial UKA, using finite-element (FE) analysis (15). Accurate in silico evaluations of knee-joint replacements are useful for clinical assessment (15).

Therefore, the objective of this study was to evaluate the biomechanical effects of medial UKA using FB and MB design prostheses, on a PF joint. The differences in the contact stress on the PF joint and the quadriceps force between the FB and MB designs were evaluated under the deep-knee-bend condition. It was hypothesized that there is no difficulty in applying medial UKA even if OA exists in the PF joint (unless there is anterior knee pain), because the differences in the biomechanical effects on the PF joint are negligible between UKA and a normal knee joint.

Methods

Normal knee model

In this study, an existing three-dimensional (3D) nonlinear FE models of the knee joint of four male subjects (Subject 1: age 36 years, height 178 cm, mass 75 kg; Subject 2: age 34 years, height 173 cm, mass 83 kg; Subject 3: age 32 years, height 182 cm, mass 79 kg; Subject 4: age 34 years, height 173 cm mass 71 kg) and one female subject (Subject 5: age 26 years, height 163 cm, mass 65 kg) was used. This FE models were developed using computed tomography and magnetic resonance imaging data (16, 17) and included the bony structures of the knee joint and the soft tissues of the PF and tibiofemoral (TF) anatomy. The articular cartilage and menisci were defined as isotropic linearly elastic materials and transversely isotropic and linearly elastic materials, respectively (18). The material properties of the articular cartilage and menisci are presented in Table 1.

Table 1
Material properties of the articular cartilage and menisci.

Cartilage	Linearly elastic, isotropic	$E = 15$ MPa $v = 0.475$
Menisci	Linearly elastic, transversely isotropic	E_θ = 150 MPa, E_r = E_z = 20 MPa v_{rz} = 0.2, $v_{r\theta}$ = $v_{z\theta}$ = 0.3, $G_{r\theta}$ = $G_{z\theta}$ = 57.7 MPa

All ligaments were modeled with nonlinear and tension-only spring elements (19, 20). The interfaces between the cartilage and bones were modeled to be fully bonded. Contact was applied between the femoral cartilage and the meniscus, the meniscus and the tibial cartilage, and the femoral cartilage and the tibial cartilage for both the medial and lateral sides (16).

Medial UKA model

An FB UKA (Zimmer, Inc., Warsaw, IN, USA) and MB UKA of the Oxford knee system (Biomet, Warsaw, IN) were virtually implanted in the medial compartment of the normal knee model. The bone models were imported and appropriately positioned, trimmed, and meshed with rigid elements according to surgical techniques (16). The tibial component was defined as a square (0°) inclination in the coronal plane with a 5° posterior slope. The rotating axis was defined as a line parallel to the lateral edge of the tibial component passing through the center of the femoral component peg. A femoral component distal cut perpendicular to the mechanical axis of the femur and parallel to the tibial cut was reproduced. The height of the PE insert was identical to the anatomy on a sagittal plane that was aligned with the mechanical axis of the tibia and positioned at the medial edge of the tibia. The materials used for the femoral component, tibial insert, tibial baseplate, and bone cement were a cobalt chromium molybdenum alloy (CoCrMo), ultrahigh-molecular-weight polyethylene (UHMWPE), a titanium alloy (Ti6Al4V), and polymethyl methacrylate (PMMA), respectively (Table 2) (17, 21, 22).

Table 2

Material properties of implant.

	Young's modulus (MPa)	Poisson's ratio
CoCr alloy	220,000	0.30
UHMWPE	685	0.47
Ti6Al4V alloy	110,000	0.30
PMMA	1,940	0.4

The femoral component and tibial insert were in contact, with a coefficient of friction of 0.04 (21). The FE simulation involved two types of loading conditions, corresponding to the loads used in the model validation experiment and to predictions of daily-activity loading scenarios. Axial loading of 1150 N was applied to the model to evaluate the contact stresses and compare them with previous studies (23). The second loading condition corresponded to a deep-knee-bend, and squat loading was applied for evaluating the knee-joint mechanics. A computational analysis was performed by utilizing an AP force applied to the femur, based on the compressive load applied to the hip with constrained femoral internal–external (IE) rotation, free medial–lateral translation, and knee flexion, for a combination of vertical hip and quadriceps loads. Therefore, a six-degree of freedom (DOF) TF joint was developed (24, 25). A proportional–integral–derivative controller was incorporated into the computational model to control the quadriceps in a manner similar to that utilized in previous experiments (26). A control system was utilized to calculate the instantaneous displacement of the quadricep muscles to match the target flexion profile used in the experiment. Furthermore, IE and varus–valgus torques were applied to the tibia, while the remaining tibial DOFs were constrained (24, 25).

The FE models were analyzed using the Abaqus software (version 6.11; Simulia, Providence, RI, USA). The contact stress and quadriceps force on the PF joint were evaluated for the FB and MB medial UKA designs.

Statistical analysis

We performed the test to divide into 11 time points (0.0 to 1.0 phases) for single cycles of deep-knee-bend loading condition. To assess the two different models, FB, and MB, each model's condition was compared to the normal knee in pairwise manner using non-parametric repeated-measure Friedman tests at each phase of the cycle. In this paper, it was used a Wilcoxon's rank test with Holm correction for the posthoc comparisons, to control the familywise error rate for the tests conducted within each phase of the cycle. Statistical analyses were performed using SPSS for Windows (version 20.0.0; SPSS Inc., Chicago, IL, USA). Statistical significance was set at $P < .05$ for all comparisons.

Results

The results of the five subject-specific FE models were compared with previous FE results for model validation (23). The average contact stresses on the medial and lateral menisci in the present study and a previous study are presented in Table 3.

Table 3
Comparison of the average contact stresses on the menisci for the validation of the model under an axial loading condition.

	Previous study [23]	Present study	Standard deviation
Medial meniscus (MPa)	2.9	3.1	0.5
Lateral meniscus (MPa)	1.4	1.5	0.4

The minor differences may be due to variations in the geometry, such as the thicknesses of the cartilage and meniscus, between the studies. However, the consistency between the results confirms the ability of the FE model to produce reasonable results (23). Figure 1 shows the contact stresses on the PF joint with the FB and MB medial UKA designs under the deep-knee-bend condition. No significant difference in the contact stress on the PF joint was observed between the FB and MB medial UKA models and an intact model at an early flexion angle. At a larger flexion angle, the results of the contact stress were showed a significant increase by 7% (on average) for the FB respectively, compared with the intact model. It is a small value but a significant increase. For the MB models, we observed an increase by 4% (on average) for contact stress. But there was no statistically significant increment.

The quadriceps forces on the PF joint for the FB and MB medial UKA designs under a deep-knee-bend are shown in Fig. 2. A larger quadriceps force was needed to produce an identical flexion angle for both the FB and MB UKA designs, compared with the intact model. The quadriceps force rapidly increased the flexion of the knee joint in all the models. On average, the maximum quadriceps force significantly ranged from 2710 N for the MB UKA design to 2830 N for the FB UKA design. At a mid-flexion angle, the quadriceps forces were smaller for the FB and MB UKA designs than for the intact model. Additionally, a lower quadriceps force was needed to produce identical flexion angles with the MB UKA design than with the FB UKA design. The FB and MB UKA designs required 12% and 8% (on average) more quadriceps force, respectively, than the intact model.

Discussion

The most important finding of this study was that the contact stress on the PF joint increased less with the MB UKA design than with the FB UKA design, however, there was no significant difference in contact stress on the PF joint between medial UKA and intact knee joints. And a lower quadriceps force was needed to produce the same flexion angle with the MB UKA design than with the FB UKA design. UKA can be performed with either an FB or an MB design. In a prospective study involving 48 patients, who were randomly assigned either FB or MB UKA prostheses, Li et al. observed better knee kinematics and a lower incidence of radiolucencies for the MB design, but the Knee Society, WOMAC, and SF-36 scores were equivalent for the two designs (27). In another study, the range of motion, limb alignment, patient-reported outcomes, incidence of aseptic loosening, and reoperation rates were identical for the FB and MB UKA designs (28). However, the time to reoperation and failure mode differed. Early failure due to bearing dislocation occurred with the MB design, whereas late failure due to polyethylene wear occurred with the FB design. A previous study indicated that during a ≥ 15-year follow-up period, some type of revision arthroplasty was required for 12 of 77 knees (15%) in the case of FB UKA (Miller–Galante; Zimmer) and for 10 of 79 knees (12%) in the case of MB UKA (Oxford; Biomet) (29). No differences were observed in the number of knees with progressive lateral OA that required revision arthroplasty, between the FB and MB UKA designs [33]. Thus, there are many arguments regarding the biomechanical issues of FB and MB UKA designs. In previous studies on the progression of OA after UKA, the radiological assessment was neither blind nor randomized (27).

The advantage of FE analysis is that the impact of UKA design can be determined without external variables [34]. Most in vitro biomechanical studies have involved evaluations using aged cadaveric subjects with loosening between the specimen and the device, as well as attenuation of the tissue, which can occur owing to successive loading in mechanical testing (26). A model of an intact joint was the foundation of this study and involved FEM validation steps. The results exhibited good agreement with those of previous computational studies (23, 30).

Therefore, the UKA models used in the present study and the related analyses are considered reliable.

Kozinn and Scott proposed that UKA should not be offered to patients with PF joint arthritis, for optimal results (31). This sparked a contentious debate on PF joint disease, because other authors demonstrated only a weak correlation between PF degenerative changes and anterior knee pain (11, 32).

Additionally, owing to differences in the design and biomechanics of FB UKA, damage to the PF joint has traditionally been a contraindication. Lim et al. recently showed that the presence of significant preoperative radiological PF disease does not affect long-term implant success, and patients had excellent postoperative functional outcomes for 10 years (33). In the present study, the MB UKA design produced a smaller increase in the amount of PF joint-contact stress compared with a model of an intact joint, than the FB UKA design produced. Previous studies on MB UKA have indicated that the presence of PF degeneration does not compromise clinical outcomes, because the implant is believed to be more patella-friendly owing to better kinematics, which supports our results (33, 34). Additionally, although the contact stress on the PF joint increased with both the FB and MB UKA designs, it was not a statistically significant value. Biomechanical studies have indicated that the progression of arthritis of a PF joint typically does not necessitate revision.

The quadriceps force needed to produce a squatting motion was greater for the FB design than for the MB design (by as much as 120 N for knee flexion angles > 100°). Thus, increased quadriceps strength leads to improved functional performance (35). Because OA and knee arthroplasty patients experience significant quadriceps weakness, the FB UKA design, which increases the required quadriceps force, can make it more difficult for patients to walk, kneel, or perform a deep-knee-bend (36). This agrees with the results of a previous in vitro study in which a UKA model required less quadriceps force at a mid-flexion angle than an intact model did (27).

From a biomechanical viewpoint, our results indicated that the risk of PF joint progressive OA can be reduced with the MB UKA design because it preserves the normal biomechanical effect, in contrast to the FB UKA design. Additionally, the MB UKA design requires a lower quadriceps force and makes it easier for recipients to kneel, squat, or rise from a chair.

Three strengths of our study should be highlighted. First, a well-validated setup that accounted for numerous previous results was employed. Second, in contrast to previous UKA studies, the tibia, femur, and related soft tissues were included in the FE model. Third, in contrast to current biomechanical UKA models, the model used in this study included the deep-knee-bend and squat loading, rather than simple vertical static-loading conditions. Despite these strengths, this study had certain limitations. First, the results do not predict clinical results or patient satisfaction. Second, the computational model was developed using data for four male subjects and one female subject. Using data for subjects of various ages would improve the validity of the results, as it would increase the diversity of the knee joint geometry. However, in this study, our objectives were to evaluate the biomechanical effect of UKA in young individuals. Third, the bony structures were assumed to be rigid. In reality, bone is composed of cortical and cancellous tissues. However, the main purpose of the study was not to evaluate the effects of different prostheses on bone. Additionally, this assumption had a minimal influence on the results of the study, because the stiffness of bone exceeds that of the relevant soft tissues (23). Finally, the simulation only involved the action of a deep-knee-bend; simulations involving rising from or sitting on chairs, climbing/descending stairs, and squatting should be performed in future investigations.

In conclusion, this study provides biomechanical evidence that degenerative changes in the PF joint should not be considered an absolute contraindication for treatment with medial UKA.

In addition, UKA is not problematic even if the PF joint has OA unless there is anterior knee pain because there was no significant difference in contact stress.

Abbreviations

UKA: Unicompartmental knee arthroplasty; OA: Osteoarthritis; PF: Patellofemoral; FB: Fixed-bearing; MB: Mobile-bearing; 3D: Three-dimensional; TF: Tibiofemoral; CoCrMo: Cobalt chromium molybdenum alloy; UHMWPE: Ultrahigh-molecular-weight polyethylene; Ti6Al4V: Titanium alloy; PMMA: Polymethyl methacry; IE: Internal–external; DOF: Degree of freedom.

Declarations

Ethics approval and consent to participate

Approval was not required, as neither human participants nor animals were involved in this study.

Consent for publication

Not applicable

Availability of data and material

Not applicable

Competing interests

The authors declare that they have no competing interests

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Author’s contributions

H.M. Kwon designed the study and drafted the paper; J.A. Lee developed the 3D model; Y.G. Koh evaluated the FEA results; K.K. Park validated the data; K.T. Kang supervised the study and analyzed the data.

Competing interests

The authors declare that they have no competing interests

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Figures

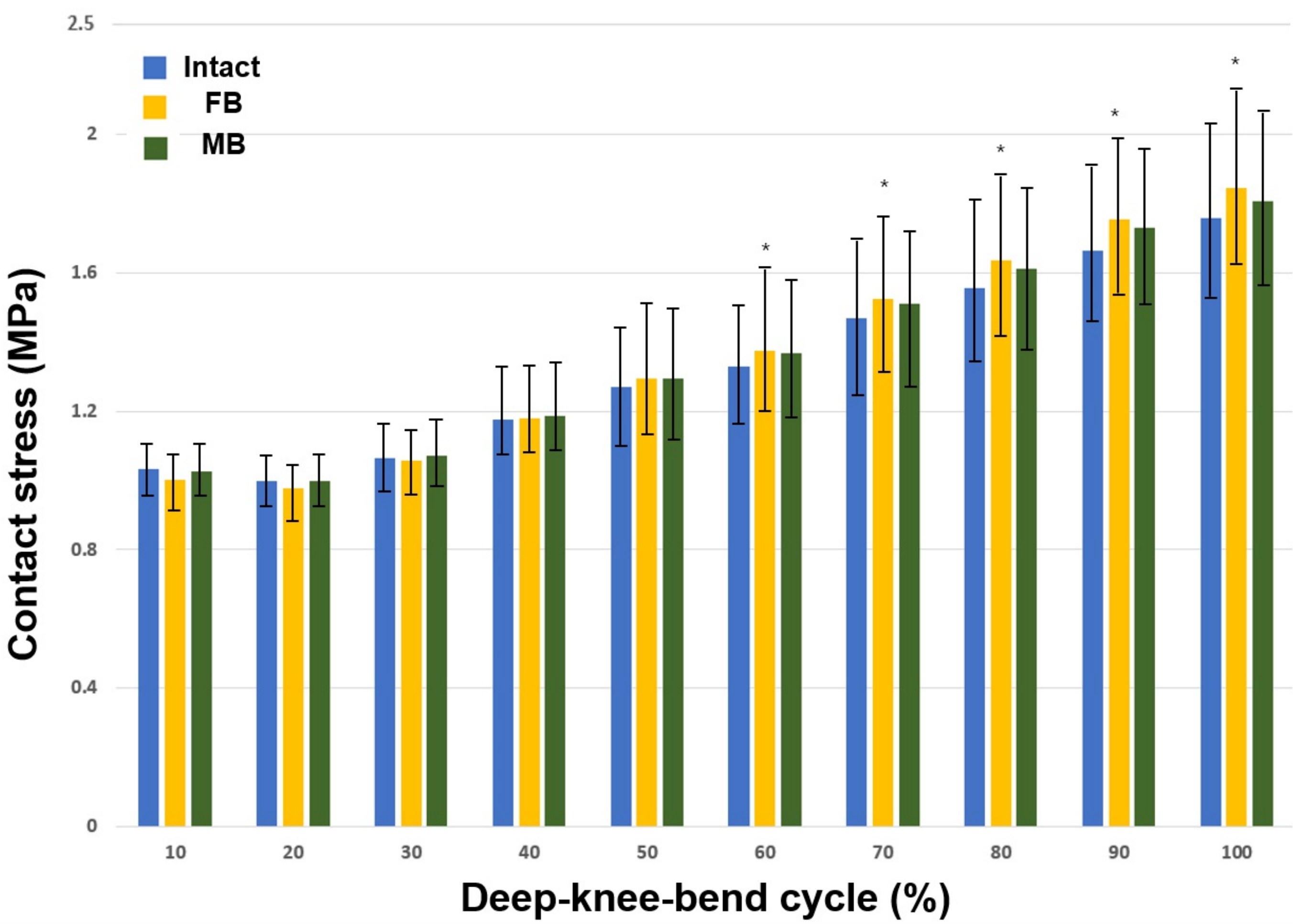

Figure 1

Differences in contact stress on the PF joint with intact, FB and MB design medial UKA under the deep-knee-bend condition (*$p<.05$).

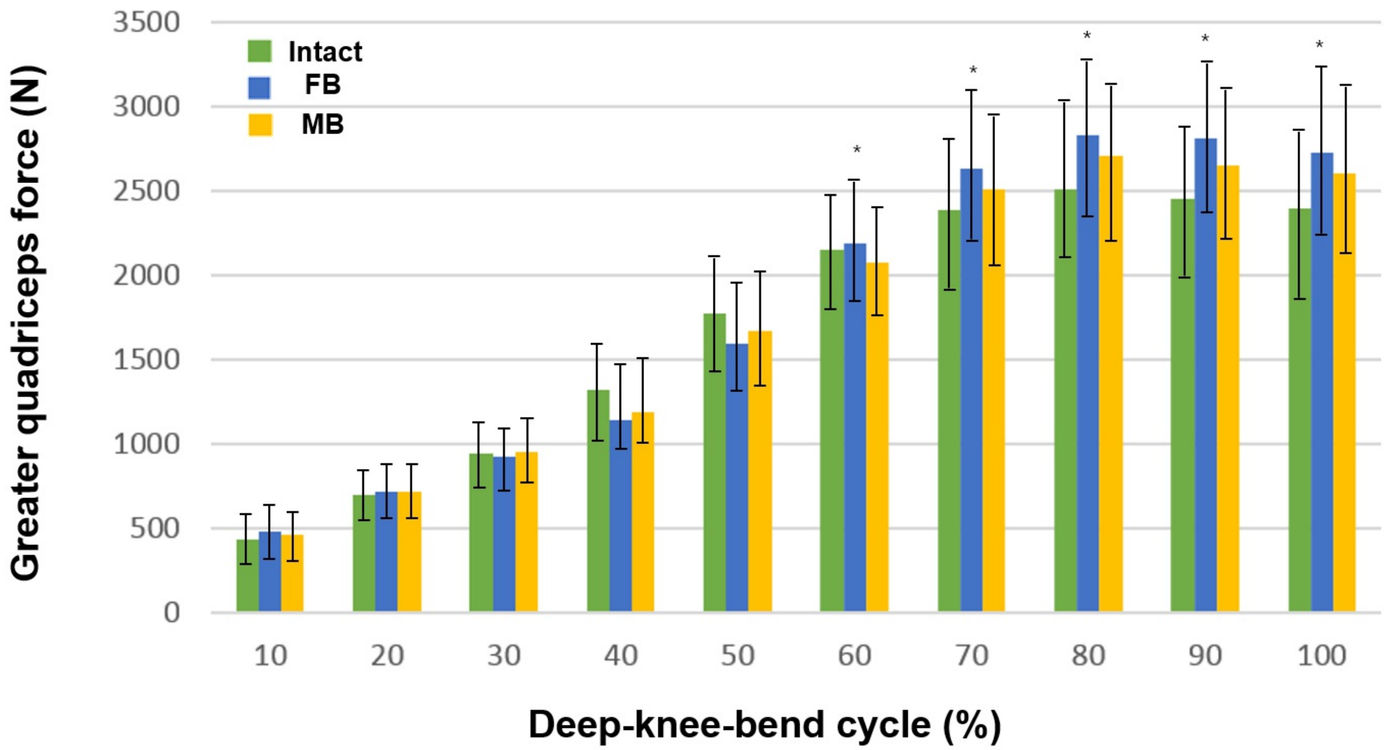

Figure 2

Differences in the quadriceps muscle force for intact, FB and MB tibial insert materials under the deep-knee-bend condition (*p<.05).